

Elderly Subjects' Ability to Recover Balance With a Single Backward Step Associates With Body Configuration at Step Contact

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Background. In the event of a slip or trip, one's ability to recover a stable upright stance by stepping should depend on (a) the configuration of the body at the instant of step contact and (b) the forces generated between the foot and ground during step contact. In this study, we tested whether these two variables associate with elderly subjects' ability to recover balance by taking a single backward step after sudden release from an inclined position.

Methods. Twenty-six community-dwelling subjects (12 women, 14 men) of mean age 75 ± 4 (*SD*) years each underwent five trials in which they were suddenly released from a backward inclination of 7° and instructed to "recover balance with a single step." Body segment motions and foot contact forces were analyzed to determine step contact times, stepping angles, body lean angles at step contact, and the magnitudes and times (after step contact) of peak foot-floor contact forces and peak sagittal-plane torques at the ankle, knee, and hip of the stepping leg.

Results. Fifty percent of subjects were predominantly single steppers (successful at recovering with a single step in greater than three of five trials), 27% were multiple steppers (successful in less than two of five trials), and 23% were mixed response steppers (successful in two of five or three of five trials). Recovery style associated with the ratio of stepping angle divided by body lean angle at step contact ($p = .003$), which averaged 1.4 ± 0.5 for single steppers and 0.6 ± 0.5 for multiple steppers, but not with step contact time, stepping angle, or contact forces and joint torques during step contact.

Conclusions. These results suggest that elderly subjects' ability to recover balance with a single backward step depends primarily on the configuration of the body (in particular, the ratio of stepping angle to body lean angle) at step contact.

STEPPING represents a primary means for balance recovery after a slip or trip (1–4). Such perturbations rank among the most common self-reported causes of falls and fall-related injuries in the elderly population (5–10). Accordingly, considerable motivation exists to develop exercise-based therapies to enhance elderly individuals' ability to recover balance by stepping.

An important prerequisite to the development of such programs is identifying the biomechanical and neuromuscular variables that govern one's ability to recover balance by stepping. However, although several studies have reported age-related declines in ability to recover balance by stepping (2,11–15), these have not revealed consistent evidence of the biomechanical variables that underlie such differences. For example, for a given perturbation strength, Luchies and colleagues (11) found that elderly women took smaller, quicker steps than young women, Wojcik and colleagues (15) found that elderly women took larger steps than young women, and McIlroy and Maki (12) found no difference in step execution times and step lengths between young and elderly subjects.

Our conceptual model of balance recovery by stepping focuses on the notion that the effective restoring moment provided by the leg during step contact largely determines whether a step is successful in restoring a stable upright

stance (16). This depends, in turn, on the position, with respect to the body's center of gravity, and force-generating capacity of the stepping leg during step contact. For example, one may be able to move the leg into a contact position that provides a large moment arm, but downward movement of the body will not be halted unless sufficient contact forces are generated between the foot and the ground, through contraction of muscles spanning the lower extremity joints. Conversely, one may be able to generate large forces between the foot and the ground during step contact, but balance recovery will not occur if the step size and moment arm are so small that such forces generate little effective restoring moment.

The aim of the current study was to determine the validity of this conceptual model among healthy elderly persons. In particular, we tested whether in the event of sudden release from a backward inclined position, elderly subjects' ability to recover balance with a single step associates with variables related to the quickness of the step, the configuration of the body at step contact, and the force-generating capacity of the stepping leg during step contact. We hypothesized that ability to recover balance with a single step would associate negatively with step execution time and positively with variables related to step size and the magnitude of foot-floor reaction forces during step contact.

METHODS

Subjects

Twenty-six community-dwelling elderly adults (12 women and 14 men) with a mean age of 75 ± 4 (*SD*) years, body height of 1.66 ± 0.11 m, and body mass of 72 ± 15 kg were recruited through advertisements in local newspapers and seniors' centers in the San Francisco Bay area. Potential subjects were excluded if they met any of the following criteria that would either prevent them from being able to perform the experiment or would represent potentially confounding (and, given our small sample size, undesirable) influences on performance: (i) inability to stand independently and walk a distance of 5 m; (ii) impairment of neuromuscular function secondary to previously diagnosed neurological disease (e.g., stroke, Parkinson's disease, peripheral neuropathy); (iii) amputations, severe arthritis, or other debilitating orthopedic problems; (iv) severely impaired vision (e.g., inability to read newsprint at arm's length with corrective lenses); (v) use of medications known to affect balance (e.g., sedatives, antiarrhythmics); and (vi) cognitive impairment (Folstein Mini-Mental State Exam score <27). Informed written consent was obtained from each subject, and the experiment was approved by both the Committee on Human Research at the University of California, San Francisco, and the Committee for the Protection of Human Subjects at the University of California, Berkeley.

Experimental Protocol

During the experiment, subjects were released suddenly from a backward inclined position by means of a horizontal tether, which attached at one end to a chest harness and at the other end to an electromagnet (Figure 1A). This protocol is similar to that used by other researchers to study balance recovery by forward stepping (13,15–18). At the initiation of each trial, the subject stood barefoot on a walkway that had a force plate of surface area 60×90 cm (model 6090H, Bertec Corp, Worthington, OH) mounted flush to its surface. A linoleum cover concealed the location of the force plate. We then instructed the subject to lean backward into the tether, the length of which was adjusted to provide an initial lean angle θ_0 of 7° from the vertical (which we determined from pilot studies to exceed sway-based recovery abilities). We then informed the subject that, in the event of tether release, he or she should “try to recover balance with a single step.” To increase the unexpectedness of the release, we randomized between 10 and 60 seconds the time delay between the instant the subject assumed the leaning position and the instant of tether release; we played music to mask equipment noise. We provided no instructions regarding whether the subject should step with the right or left leg. The subject placed his or her arms initially to the side and directed his or her gaze forward. For safety, the subject was secured to a second fall restraint harness (which was slack during stepping), and members of the research team were positioned nearby as “spotters.”

Following three “practice” trials (which familiarized the subject with the experimental protocol and allowed us to discreetly position the subject so the first step landed on the force plate), five “actual” trials were acquired. During the

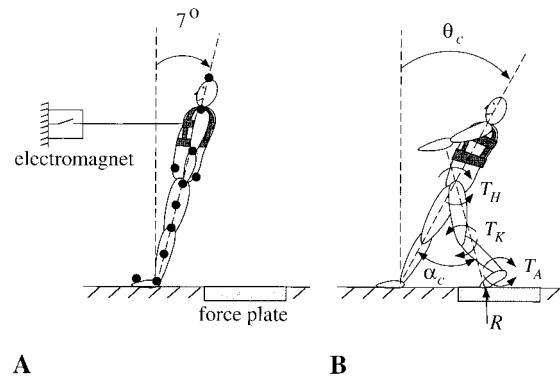


Figure 1. Balance recovery by stepping experiment. **A**, A horizontal tether and electromagnet were used to release subjects from a backward inclination. Body segment motions were determined based on the positions of 20 skin-surface markers located at the crown of the head and L5-sacral junction, and the right and left acromion, lateral humeral epicondyle, distal intersection of the radius and ulna, junction of the second and third metatarsal, lateral malleolus, mid-shin, lateral femoral epicondyle, mid thigh, and anterior superior iliac spines; **B**, in each trial, a six-camera, 60-Hz motion analysis system recorded body segment positions and a force plate measured foot contact forces (R). The stepping angle (α_c) and body lean angle (θ_c) were calculated based on body configuration at toe contact. Joint torques due to muscle contraction at the ankle, knee, and hip (T_A , T_K , T_H) during step contact were calculated from inverse dynamics. Joint rotation was defined positive for dorsiflexion at the ankle and for flexion at the knee and hip. Joint torque was defined positive if plantar flexor at the ankle and extensor at the knee and hip. Ankle and knee joint centers were estimated from markers overlying the lateral malleolus and femoral epicondyle, respectively. The hip joint center was estimated from the anterior superior iliac spine and sacral markers, using a routine developed by Vaughan and colleagues (19).

latter, synchronized recordings were acquired via the force plate and a six-camera, 60-Hz motion capture system (MacReflex, Qualisys Inc, Glastonbury, CT) of the contact force generated between the foot and the ground and the three-dimensional positions of 20 surface markers placed bilaterally (Figure 1A). The MacReflex system has a measurement accuracy of approximately 1.0 mm, which was more than adequate for the large motions associated with the experiment. Marker position data were filtered with a recursive, fourth-order low-pass Butterworth filter, with a 6-Hz cutoff frequency. From each trial, we determined the step contact time, the body lean angle at step contact, the stepping angle at step contact, and the ratio of stepping angle divided by body lean angle (16). Only the first step of each trial was analyzed. The step contact time (t_{step}) was calculated as the time interval, in milliseconds, between tether release and initial contact of the stepping foot with the force plate. The body lean angle at step contact (θ_c) was determined as the sagittal plane projection of the angle, in degrees, between the vertical and the body lean axis, which we defined by the line connecting the ankle of the stance (or pivot) foot to the midshoulder position (Figure 1B). The stepping angle (α_c) was determined as the sagittal-plane projection of the angle, in degrees, between the body lean axis and the stepping leg axis, where the latter was defined by the line connecting the toe of the stepping leg and midpoint of the pelvis.

In general, θ_c influences the moment arm of the gravitational destabilizing force at the instant of step contact, α_c influences the moment arm of the stepping leg, and α_c/θ_c reflects the mechanical efficiency of the step. More specifically, the gravitational moment that must be overcome to halt downward rotation of the body increases with increasing θ_c , increasing body height, and increasing body mass. (Of course, inertial moments due to angular accelerations of the body segments must also be overcome, but for a given initial lean angle θ_o , these should also increase with increasing θ_c .) Furthermore, the moment arm of the stepping leg increases with increasing α_c , assuming that foot contact forces are directed approximately along the stepping leg axis (an assumption that appears to be justified from analysis of our data) and α_c is less than 90° . The ratio of the restoring moment to the destabilizing moment at the instant of step contact should therefore associate with body mass, body height, and the ratio of stepping angle divided by body lean angle (α_c/θ_c). Accordingly, α_c/θ_c can be interpreted to reflect the mechanical (or kinematic) efficiency of the step (16).

We also estimated temporal variations in torques at the hip, knee, and ankle of the stepping leg during step contact, from inverse dynamics analysis of body segment motions and foot contact forces (19,20). This technique idealizes the body as a chain of rigid links (having inertial properties scaled to body height and weight) connected by frictionless joints. The net torque due to contraction of muscles spanning each joint (from the ankle upward) is calculated from Newton's equations of motion, based on temporal variations in linear and angular positions, velocities, and acceleration of individual body segments, the magnitude of foot contact force, and the point of application of the resultant force along the base of the foot (center of pressure). We determined both the magnitude and time (after step contact) to peak ankle plantar flexor, knee extensor, and hip flexor torques, which were the primary stabilizers of the leg and trunk during step contact (Figure 2C). Following standard conventions, peak torques were normalized by the product of body mass times body height.

Data Analysis

Subjects were classified into three recovery styles based on their ability to perform the specified task of recovering balance with a single step. A trial was defined as a "single step" recovery if only one step was taken or if the length of a second step (of the contralateral leg) did not exceed that of the original step. A "single stepper" was defined as a subject who recovered balance with a single step in four or five trials. A "multiple stepper" was defined as a subject who recovered balance with a single step in one or no trials. A "mixed response" subject was defined as a subject who recovered balance with a single step in two or three trials. Three subjects required support from the fall restraint harness to recover balance in one or more trials. Such trials all involved more than one step and were therefore classified as multiple step trials.

Based on the above criteria, 50% of subjects ($n = 13$) were classified as single steppers, 27% ($n = 7$) were multiple steppers, and 23% ($n = 6$) were mixed response. Sixty-two percent of single steppers ($n = 8$) used one step to recover

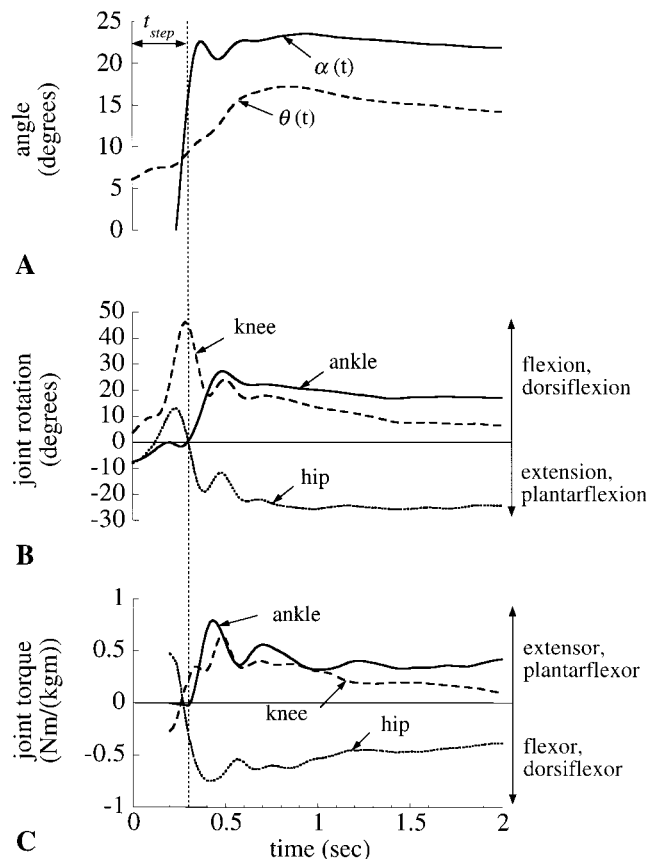


Figure 2. Temporal variations in step kinematics and kinetics during successful balance recovery by stepping. Tether release occurred at $t = 0$, and toe-off occurred at approximately 200 ms. The dashed vertical line shows the instant of step contact. **A**, Step contact was achieved approximately 300 ms after tether release, with a corresponding α_c of 15° , θ_c of 9° , and α_c/θ_c of 1.7. **B**, Immediately following step contact, the ankle rotates into increasing dorsiflexion, and the knee and hip rotate into increasing extension. Note that hip rotation is defined by the angle in the sagittal plane between the trunk and thigh of the stepping leg. **C**, Joint torques due to muscle contraction are initially eccentric or energy absorbing at the ankle and hip, and concentric or energy generating at the knee. Torques were normalized by the product of the subject's body mass and body height (67 kg and 1.65 m) and were calculated only for the interval beyond the instant of toe-off.

balance in all five trials, and 83% of multiple steppers ($n = 5$) used multiple steps to recover balance in all five trials.

One-way analysis of variance (ANOVA) was used to assess whether recovery style associated with average values (over the five repeated trials) of t_{step} , α_c , α_c/θ_c and the times and magnitudes of peak contact forces and joint torques. Foot contact forces and joint torques in the stepping leg could not be determined for four subjects, because of a malfunction in acquiring data from the force plate. Accordingly, only kinematic data (t_{step} , α_c , α_c/θ_c) for these subjects were included in the analysis. Post hoc multiple comparison tests (Tukey honestly significant difference) were performed if an ANOVA revealed a significant association. Chi-squared analysis was used to test for association between recovery style and gender. Associations between continuous variables were examined through correlation. All statistical

tests were performed with the SPSS statistical software package (SPSS, Inc, Chicago, IL). Based on a Bonferroni correction for multiple statistical comparisons, we regarded p values less than .005 to indicate a significant association, and p values between .05 and .005 to reflect a “borderline” association.

RESULTS

Recovery style associated significantly with α_c/θ_c ($p = .003$; Table 1 and Figure 3). Post hoc tests indicated that significant differences existed in this variable between single steppers (average value: 1.4 ± 0.5 [SD]) and multiple steppers (average value: 0.6 ± 0.5), but not between single steppers and mixed response, or between multiple steppers and mixed response. In contrast, no association existed between recovery style and t_{step} , α_c , or the times and magnitudes of peak forces and joint torques (Tables 1 and 2). Borderline association ($p = .04$) existed between gender and recovery style, with 10 of the 14 male subjects, as opposed to 3 of the 12 female subjects, being single steppers.

Association existed between α_c/θ_c and α_c ($r = .92, p < .0001$), but not between α_c/θ_c and θ_c ($r = .13, p = .51$), t_{step} ($r = .46, p = .017$), peak contact force ($r = .22, p = .39$), or gender ($p = .18$). Borderline associations were observed between α_c and t_{step} ($r = .51, p = .007$) and between α_c and θ_c ($r = .50, p = .010$), suggesting that larger steps tended to take longer to execute and involved greater body lean angles at step contact.

DISCUSSION

We found that recovery style associated significantly with the ratio of stepping angle divided by body lean angle at step contact (α_c/θ_c), which was larger for single steppers than multiple steppers. The strength of this parameter in separating single and multiple steppers likely relates to its ability to summarize the combined influence on recovery style of step execution time and stepping angle, which in themselves were weaker indicators of performance.

We observed no association between recovery style and the magnitude or timing of foot-floor contact forces (or lower extremity joint torques) during step contact. From a biomechanical perspective, one should be able to compensate for inefficient positioning of the stepping foot (i.e., small α_c/θ_c) by generating greater foot contact force (16). However, we observed no correlation between α_c/θ_c and peak contact force. This suggests the possibility that observed contact

forces (which averaged 133% of body weight in multiple steppers and 126% in single steppers) represented near-maximal efforts, which resulted in balance recovery only when associated with efficient positioning of the stepping leg. However, given our experimental design, we have no means for confirming that single steppers approached their true force-generating capacity or that the latter parameter is not affected by leg configuration during step contact. Such uncertainties indicate the need for additional, more controlled experiments to determine the relations between α_c/θ_c , force generating capacity, and recovery style.

Our results are in general agreement with those of previous studies on balance recovery by stepping. Wojcik and coworkers (15) found that elderly men were more able than elderly women to recover balance with a single step after release from a forward inclination. Our results suggest that a similar trend exists for release from an initial backward inclination. McIlroy and Maki (12) combined data from stepping experiments with young and elderly subjects and found that single steppers used larger steps and longer step execution times than multiple steppers. In our study, neither of these parameters associated with recovery style. However, we found that increases in step size associated with increased α_c/θ_c (which in turn associated with recovery style) and that larger steps tended to associate with increased delays in step execution time. Luchies (21) observed no association between recovery style and lower extremity joint torques during step initiation. Our study cannot confirm this result, because we did not examine joint torques during the initiation and swing phases of stepping.

Several limitations exist to this study. We included only healthy elderly subjects, and quite different trends might be observed among frail elderly persons. We also examined only backward perturbations to balance, and thus, additional studies are required to test whether our findings apply to forward and sideways perturbations. Although sideways falls are associated with a higher risk for hip fracture, backward falls are also the cause of significant morbidity, including approximately 17% of hip fractures and 26% of wrist fractures (22,23).

Furthermore, we instructed our subjects to “try to recover balance with a single step,” and one might argue that a more general instruction (such as “recover balance” [17] or “prevent a fall” [12]) better represents one’s goal during a real-life loss of balance. One might also argue that the use of multiple steps to recover balance following a real-life per-

Table 1. ANOVA Results on Association Between Step Characteristics and Recovery Style

Parameter	Recovery Style			p Value
	Single Steppers (n = 13)	Mixed Response (n = 6)	Multiple Steppers (n = 7)	
Step execution time t_{step} (ms)	350 ± 50	380 ± 60	350 ± 60	.6
Stepping angle α_c (°)	15 ± 7	13 ± 2	9 ± 8	.1
Mechanical advantage α_c/θ_c	1.4 ± 0.5	1.1 ± 0.1	0.6 ± 0.5	.003
Body lean angle at contact θ_c (°)	11 ± 2	12 ± 2	13 ± 2	.09
Peak contact force/body weight	1.26 ± 0.17	1.29 ± 0.04	1.33 ± 0.12	.7
Time to peak contact force (ms)	170 ± 50	170 ± 60	150 ± 20	.6

Notes: Cell entries show mean ± 1 SD; ANOVA = one-way analysis of variance.

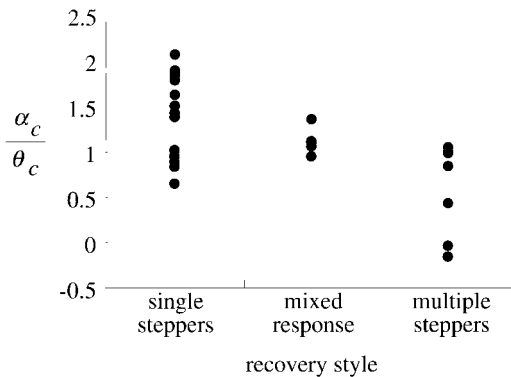


Figure 3. Balance recovery style associated with α_c/θ_c with single steppers exhibiting higher magnitudes of this ratio than multiple steppers.

turbation represents a conservative strategy that helps to ensure balance recovery. We do not disagree with these possibilities. Our goal in the present study was not to assess our participants' risk for falls or to determine the relative merits of single step versus multiple step recovery styles. Instead, it was to identify the biomechanical variables that associate with elderly subjects' ability to perform a specific stepping-based balance recovery task.

We perceived that if subjects had been instructed to simply prevent a fall, difficulties would arise in the following: (a) quantifying their ability to perform this task, given safety constraints on eliciting falls in elderly subjects; and (b) identifying the relative importance to observed responses of behavioral variables (e.g., habit or preference) versus biomechanical variables (e.g., step size, step execution time). By following other investigators' approach of instructing subjects to recover balance with a single step (13,15), little risk for injury was presented to our subjects, and a rational means existed for quantifying subjects' behavior (i.e., based on whether they were successful in performing this task). Furthermore, the explicit nature of the task likely facilitated our goal of identifying biomechanical as opposed to behavioral influences on performance.

Regardless of whether a single step or multiple step strategy is used, we see little reason for promoting the use of

small (and thus biomechanically inefficient) steps. Two of the three subjects who experienced falls during our trials (i.e., required support from the fall restraint harness) attempted to recover balance with multiple small steps. For both of these subjects, the first step was so small that the toes of the stepping foot did not move behind the heel of the stance foot, and subsequent steps were not much larger. This created a situation where, despite the use of multiple steps, the foot base of support could not "catch up" to the backward-moving center of gravity of the body (a situation commonly used to describe the cause of falls in individuals with Parkinson's disease). Such considerations argue toward the use of large steps, regardless of whether recovery occurs within one or more steps.

In summary, we found that elderly subjects' ability to recover balance with a single backward step associated with the ratio of stepping angle divided by body lean angle at step contact. This ratio associated with stepping angle, but not with step execution time. In our opinion, these results suggest that regardless of whether exercise-based therapies for fall prevention focus on training single step or multiple step balance recovery, they should condition subjects to achieve large, thus biomechanically efficient, steps and target those components of lower extremity strength, flexibility, and reaction time that govern this capacity.

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Table 2. Magnitudes and Times to Peak Joint Torques During Step Contact

Parameter	Mean Value (n = 22)
Ankle	
Peak plantar flexor torque (Nm/kgm) [†]	0.57 ± 0.17
Time to peak torque (ms)	130 ± 40
Knee	
Peak extensor torque (Nm/kgm) [†]	0.61 ± 0.17
Time to peak torque (ms)	160 ± 90
Hip	
Peak flexor torque (Nm/kgm) [†]	0.65 ± 0.21
Time to peak torque (ms)	100 ± 50

Note: Cell entries show mean ± 1 standard deviation.

[†]For each subject, peak torque magnitudes (in Nm) were normalized by the product of body mass (in kg) and body height (in m).

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